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I, KIM MARSHALL, MANAGER PATENT OPERATIONS hereby certify that annexed is a true copy of the Provisional specification in connection with Application No. PP 7185 for a patent by THE LIONS EYE INSTITUTE OF WESTERN AUSTRALIA INCORPORATED filed on 18 November 1998.

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PROVISIONAL SPECIFICATION

Applicant(s):

THE LIONS EYE INSTITUTE OF WESTERN AUSTRALIA INCORPORATED

Invention Title:

LIMITED COHERENCE STEREO OPHTHALMOSCOPE

The invention is described in the following statement:

directional, scanning mechanisms. Light from the laser is reflected off the retinal wall towards a photosensitive detection device. Electro-optical circuitry is employed to convert the light into synchronized signals, so that it is possible to display an image of the fundus on a television screen or monitor.

However, although the optic disc region and retinal layers have a three dimensional structure, the existing SLO technology described above does not permit stereoscopic viewing of the ocular fundus. Stereoscopic images of the ocular fundus can impart valuable information that cannot otherwise be derived from a two dimensional representation, especially in relation to the diagnosis of glaucoma. Efforts have therefore been made to create a device capable of producing three dimensional fundus images, while improving on the contrast and resolution of conventional SLO images.

Frambach, Dacey and Sadun (1992, 1993) describe a method of producing a three dimensional fundus picture during fluorescein angiography, using a modified SLO. To obtain stereoscopic data the SLO was manually moved from side to side during angiogram proceedings, much like a fundus camera is moved to enable viewing from two different positions. Individual frames from the video tape were chosen from left and right perspectives to provide a three dimensional image. An alternative approach employed by the Frambach *et al.* involved the use of a modified Allen separator. A piece of flat glass was attached to an extended rod, which was coupled to the Allen separator, so that the glass was interposed between the eye and the SLO. The glass was then rapidly rotated to provide the left and right perspectives. The resulting frames were digitized by computer and viewed directly on a video screen. Superimposed images were formed by breaking a stereo pair down into corresponding fields and recombining them to form

the retina. By adjusting the focal plane of the aperture, images can be captured at different levels in the fundus, to reproduce desired depth characteristics. In this way a number of "optical sections" can be produced. A computer can then be used to extract depth information, through the process of "stacking" a selection of the optical sections taken at different levels of the retina. Information regarding the third dimension can therefore be interpolated.

US Patent No. 4,900,144 (also see *Optics Communications*: 87(1,2): 9-14) describes a scanning laser ophthalmoscope that employs an alternative confocal focussing arrangement. The invention can produce a three dimensional representation of an object that displays multiple reflectivity characteristics (such as the ocular fundus) through a method slightly different from the conventional confocal depth production methods described above. This US patent teaches the use of two separate confocal slit apertures and photodetecting units. The detection slits are orientated parallel to the direction in which the light, reflected from fundus, is scanned. However, both slits are slightly displaced from the normal position: the apertures are not conjugate with the fundus of the eye. One is positioned slightly forward of the conjugate plane, while the other is placed to the rear. Owing to the positioning of the confocal apertures, the output signals from the photodetectors vary in intensity according to the unevenness of the fundus. The resulting output signals are processed electronically by division calculations, detailed in US Patent No 4,900,144, to obtain a three dimensional profile. The resultant real-time image displays the topography of the fundus through different shade levels, reflecting different retinal depth levels. Software may also be used to create three dimensional graphic patterns.

The methods described above use the depth discrimination,

on the focussing or imaging qualities of optical elements as do, for example, the cornea and the lens of the eye. Interference patterns can only occur when the difference between the length of the reference arm and the object arm is shorter than the coherence length of the light source. Non-laser light sources such as bulbs, LEDs or superluminescent diodes have a coherence length of only a few micrometres. Thus, the detection of interference patterns means that the object distance is equal to the reference distance determined with an accuracy equal to the coherence length.

Popoleanu *et al.* (*Journal of Biomedical Optics*, Jan. 1998, vol. 3, no. 1, p. 12-20) describe an apparatus suited for transversal and longitudinal imaging of the retina using low coherence reflectometry. The light in the object arm of a fibre-based interferometer, where a superluminescent diode is the light source, propagates through a phase modulator and is scanned over the retina in a raster pattern. The light reflected from the fundus of the eye is combined with the light in the reference arm of the interferometer which length is controlled by means of moveable mirrors. Frequency sensitive detection recognizes the occurrence of interference. The strength of the interference signal is used by a frame grabber to form an image of features the are situated in a thin layer in the back of the eye. The axial position of this layer is controlled by the movable mirrors in the reference arm while the thickness of the layer is determined by the coherence length of the light source. In much the same way as with a confocal scanning laser ophthalmoscope a three dimensional image can be computed from a number of optical sections (transversal images) at different axial positions. Alternatively, scans in only one lateral and the axial direction (x-z-scans) may be carried out to record longitudinal images. The images acquired with this apparatus have a high depth resolution, about ten times

illuminating beam into a focused and a collimated component the Baumgartner *et al.* use a special Fresnel lens-like diffractive optical element which is placed in front of the eye. However, the use of this non-standard element can cause unwanted back reflections and transmission losses. Lateral scanning of the beam, which is necessary to acquire data not only for one point but for a line or an area of the retina, is also severely restricted.

In order to detect the occurrence of interference a time variation in the interference pattern is necessary. In this lay-out the Doppler-shift caused by the moving reference mirror creates this time variation. It is therefore not possible to record a limited coherence image of a layer of the retina for a fixed axial position of this layer.

Accordingly, there remains a need to provide a limited coherence scanning ophthalmoscope capable of acquiring images with a high depth resolution and of quantifying the 3D-morphology of the back of the eye, which is not restricted by any of the aforementioned limitations.

It is therefore an object of the present invention to provide an improved method and apparatus for producing a high contrast, three dimensional representation of a scanned object, based on both the high lateral resolution of the reflection characteristics of that object and the high axial resolution of the limited coherence reflectometry.

It is a further object of the present invention to achieve the above object with a novel limited coherence scanning ophthalmoscope design that incorporates the use of additional beam splitting, focussing and beam combining components to gain information and quantitative topographic data about the third dimension.

Preferably said apparatus includes interferometric means including said first beamsplitter and two mirrors wherein at least one of the mirrors is movable and position controllable.

Preferably said apparatus includes modulation means for modulating at least one of the said first and second beams.

Preferably the imaging apparatus includes a scanning laser ophthalmoscope including said focussing means, first and second beam scanners and said beam steerer.

Preferably said beam source includes a light source.

Preferably said apparatus includes polarisation influencing means for controlling and altering the polarisation of said first and second beams.

Preferably said light source is one of a plurality of light sources.

The apparatus may includes reflecting means, and more preferably a curved mirror, for directing light onto said surface.

The apparatus may include an aperture located optically upstream of said photodetector.

Preferably said apparatus includes a plurality of mirrors for directing said beam onto said first scanning means.

Preferably said polarisation influencing means includes polarisers, waveplates, prisms and/or beam splitting cubes.

Preferably said first direction is perpendicular to said second direction.

acousto-optic deflector.

Preferably said second scanning means is a mirror mounted on a scanning galvanometer motor.

Preferably said beam steerer is a pair of toggling mirrors that toggle every alternate frame or half frame to image the surface from two different positions, with substantially overlapping imaging areas, such that a triangulation base can be created.

Preferably said reflecting means is a large, curved mirror.

Preferably said aperture means include an iris, slit, diaphragm or a pinhole with a size to transmit single interference fringes.

Preferably said photodetector is a photomultiplier tube or an avalanche photodiode.

Preferably said signal processing means is a computer with a video signal capture facility.

Preferably said display means is a computer monitor or any other suitable display apparatus.

Preferably said apparatus includes or is couplable to signal processing means and display means for processing and displaying a resultant image of said surface.

Preferably said apparatus includes imaging analyzing means to obtain three dimensional topological data of said surface.

Preferably said apparatus includes a scanning laser ophthalmoscope.

image can be viewed and three dimensional topological data of said surface can be obtained.

Preferably said method includes directing said beam onto said first and second scanning means by means including a plurality of mirrors.

Preferably said polarisation influencing means includes polarisers, waveplates, prisms and beam splitting cubes.

Preferably said first and second directions are perpendicular to one another.

Preferably one of said first and second directions is horizontal and the other vertical.

Preferably said two different positions from which said beam is impinged onto said surface are left and right positions.

Preferably said surface is the ocular fundus.

Preferably said focussing means includes a plurality of movable and/or fixed curved and flat mirrors and/or lenses.

Preferably said light source is a superluminescent diode, a light emitting diode or a filament based bulb.

Preferably said light source is one or more of a plurality of light sources.

Preferably said beamsplitter includes an optical medium which reflects a fraction of the incident light and transmits the remainder.

Alternatively said beamsplitter includes an optical medium which reflects light of a certain polarization direction

diaphragm or a pinhole with a size to transmit single interference fringes.

Preferably said detecting means is a photomultiplier tube or an avalanche photodiode.

Preferably said signal processing means is a computer with a video signal capture facility.

Preferably said display means is a computer monitor or any other suitable display apparatus.

Preferably said image analyzing means includes computer software that recognizes image features and carries out length measurements.

According to a third broad aspect of the present invention there is provided an apparatus for visualising the ocular fundus of an eye and providing three dimensional topological data of said fundus including:

a light source for producing a beam of short coherence length;

an interferometer for dividing said beam into sub-components with a defined path difference;

modulation means for modulating at least one of the said sub-components;

beam shaping means for shaping said beam and/or said sub-components;

polarisation influencing means for controlling and changing the polarisation state of said beam and/or said sub-components;

a first beamsplitter and recombining means for splitting and re-combing said sub-components;

first focussing means for focussing said sub-components;

a second beamsplitter for splitting the sub-components;

The modulation means may include an electro-optic phase modulator.

Alternatively said modulation means may include a fibre based phase modulator.

Preferably said beam shaping means includes a beam collimator and/or expander.

Preferably said polarisation influencing means includes polarisers, waveplates, prisms and beam splitting cubes.

Preferably said beamsplitter and/or re-combining means includes an optical medium which reflects a fraction of the incident light and transmits the remainder.

Alternatively said beamsplitter and/or re-combining means includes an optical medium which reflects light of a certain polarization direction and transmits the remainder.

The beamsplitter and/or re-combining means may be coated or non-coated.

Preferably said focussing means includes a plurality of movable and/or fixed curved and flat mirrors and/or lenses.

Preferably said first scanning means includes a mirror on a resonant scanner.

Alternatively said first scanning means includes a rotating polygon mirror or a mirror on a galvanometer motor.

Preferably said second scanning means is a mirror mounted on a scanning galvanometer motor.

Preferably said reflecting means is a large, curved mirror.

a light beam 2. Alternatively, two or more light sources may be utilised to produce beam 2. This beam is directed through a polariser 3 onto a first beamsplitter 4 which forms part of an interferometric set-up. Components 5 and 6 of beam 2 are impinged onto mirrors 8 and 9 respectively, and reflected back to first beamsplitter 4. Mirror 8 is movable along the axial direction and the position of mirror 8 is computer controlled. Component 6 passes through the high fixed frequency phase modulator 7 which continuously changes the phase but not the intensity or the polarisation state of component 6. Alternatively, the Doppler effect may be used to create a frequency shift in component 6 when moving the mirror 8. Beamsplitter 4 re-combines components 5 and 6 to form beam 10 which consists of two sub-components with a difference in path length determined by the positions of mirrors 8 and 9. The beam 10 passes through half-wave plate 11, which rotates the polarisation direction of the linearly polarised beam 10 and determines the relative intensities of the components 14 and 15. Alternatively, a variable retarder in conjunction with a quarter wave plate can replace half wave plate 11. Beam 10 is then guided (through beamsplitter 12-see below) onto a second beamsplitter in the form of polarizing beamsplitter 13, which produces the two components 14 and 15 with perpendicular polarisation directions. Component 14 is impinged off mirror 17 towards lens 18 to produce a non-collimated beam in such a way that component 14 will finally be focused on the cornea of the eye 25. Alternatively, a combination of lenses and/or flat and/or curved mirrors can replace lens 18. Component 15 passes through beamsplitter 16, and a portion of component 15 reaches the beamsplitter 19, preferably a Thompson prism. Beamsplitter 19 re-combines components 14 and 15 to form beam 20.

The beam 20 is directed towards the focussing unit 21, consisting of a lens or combinations of lenses and mirrors.

component reflected from the fundus. The latter component reaches the non-polarising beamsplitter 16 where parts of the light reflected from the fundus form beam 27. The intensity of beam 27 is measured by photodetector 28, preferably a photomultiplier tube or an avalanche photodiode, to form the standard scanning ophthalmoscopic image of the fundus of the eye. Alternatively, focussing optics and/or an aperture may be placed in front of the photodetector, or optional photodetector 32 and/or 33 may be used to detect the standard scanning ophthalmoscopic image which is offset by the reflected intensity of the cornea. Alternatively, half wave plate 11 may be adjusted so that component 14 vanishes and photodetector 32 and/or 33 detects the standard scanning ophthalmoscopic image. Beamsplitter 13 re-combines the light which passes through beamsplitter 16 with light which leaves beamsplitter 19, passes through lens 18 and is reflected off mirror 17 to form beam 26.

Upon reaching the third beamsplitter in the form of non-polarizing beamsplitter 12, part of the light is reflected by the beamsplitter towards the polarising beamsplitter 29 which is rotated by 45°. Both the beams 30 and 31, which are polarised perpendicular to each other, contain reflected light from both the cornea and the fundus of the eye 25. Interference in both the beams 31 and 32 will occur when the path difference of the light reflected from the cornea and the light reflected from the fundus equals the optical distance between the cornea and the fundus within the tolerances of the coherence length of the light source 1. Since the path difference of the two sub-components of beams 10 and 20 is set by the positions of the mirrors 8 and 9, the distance from the fundus feature which is imaged to the cornea can be measured by determining the positions of the mirrors 8 and 9 when interference is established.

overlaid onto a standard ophthalmoscopic image are recorded. After storing the image the path difference between the sub-components of the illumination beam is changed by a certain amount (step width) and images at the positions A and B are acquired in the same manner as described above. The process of changing the path difference between the sub-components of the illumination beam and the acquisition of images at the positions A and B continues until the full three dimensional region of interest of the fundus of the eye is covered.

In order to determine the distance of the feature D from the reference line AB computer software is used to detect those images in the stacks of images recorded from positions A and B respectively where the feature D is marked as causing interference. The path difference between the sub-components associated with these images is a measure of the distances AD and BD, respectively. The distance AB is controlled by the positions of certain optical elements of the apparatus shown in Figure 1 and therefore also known. As a result, the triangle ABD is completely determined without measuring any angles. Triangulation methods can now be used to calculate the length of any line, including the height h (the height of triangle ABD along CD) and the value of any angle of this triangle.

Further information such as the value of the refractive index can be obtained from angular measurements. The nodal point of the scanning pattern may be placed at position C on the cornea, halfway between positions A and B. The horizontal scan angle α under which the feature D is imaged corresponds to a certain position of the horizontal scanner of the apparatus shown in Figure 1 and can be detected. The angle β, which describes the direction a beam has to travel from position C to reach feature D, can be calculated using trigonometric methods. The refractive

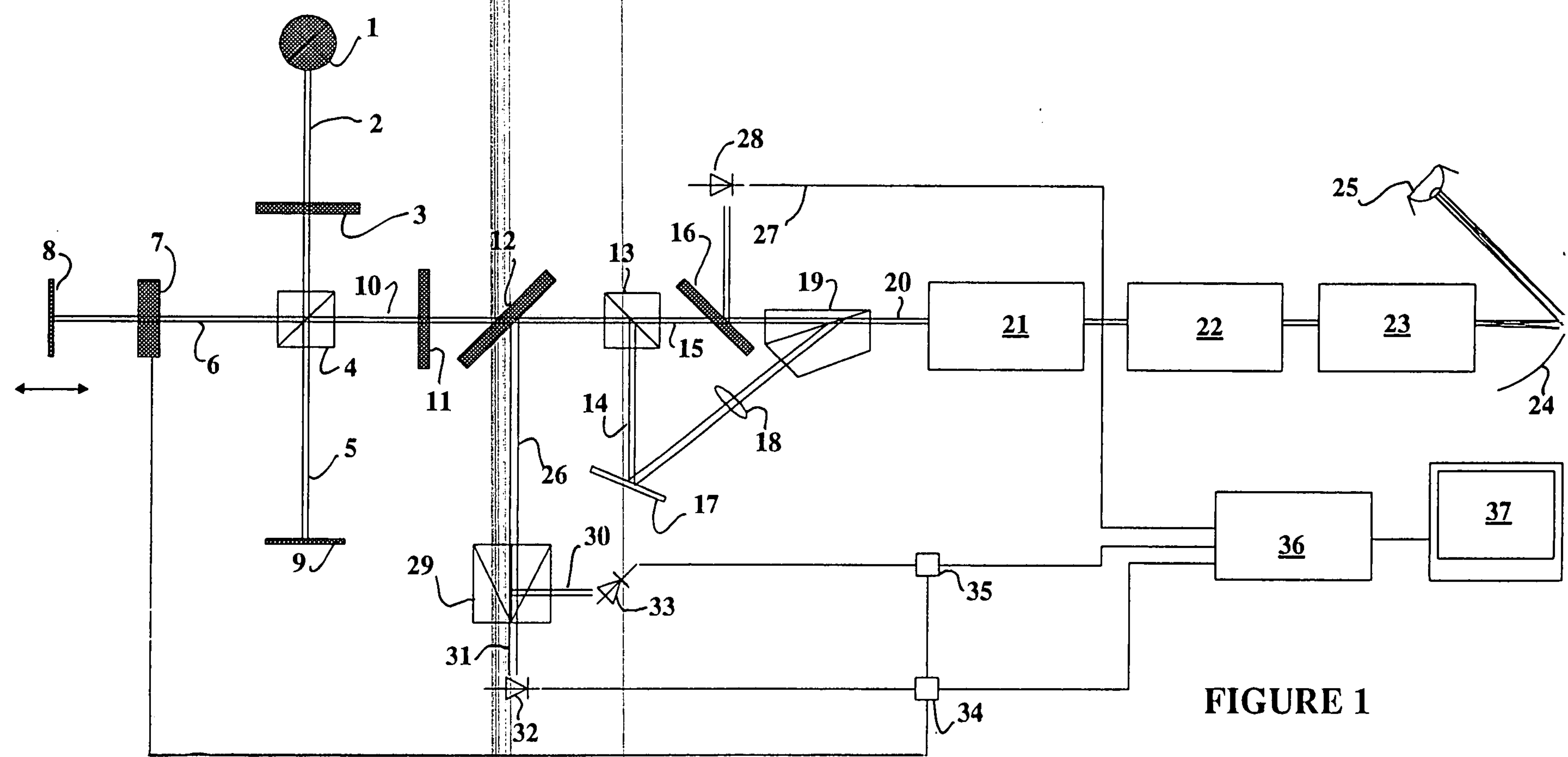
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FIGURE 1